Intraoperative cone-beam CT for correction of periaxial malrotation of the femoral shaft: A surface-matching approach

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Limb length, alignment and rotation can be difficult to determine in femoral shaft fractures. Shaft axis rotation is particularly difficult to assess intraoperatively. Femoral malpositioning can cause deformity, pain and secondary degenerative joint damage. The aim of this study is to develop an intraoperative method based on cone-beam computed tomography (CBCT) to guide alignment of femoral shaft fractures. We hypothesize that bone surface matching can predict malrotation even with severe comminution. A cadaveric femur was imaged at 16 femoral periaxial malrotations (−51.2° to 60.1°). The images were processed resulting in an unwrapped bone surface plot consisting of a pattern of ridges and valleys. Fracture gaps were simulated by removing midline CT slices. The gaps were reconstituted by extrapolating the existing proximal and distal fragments to the midline of the fracture. The two bone surfaces were then shifted to align bony features. Periaxial malrotation was accurately assessed using surface matching ($r^2=0.99$, slope 1.0). The largest mean error was 2.20° and the average difference between repeated measurements was 0.49°. CBCT can provide intraoperative high-resolution images with a large field of view. This quality of imaging enables surface matching algorithms to be utilized even with large areas of comminution. © 2007 American Association of Physicists in Medicine. [DOI: 10.1118/1.2710330]

Key words: femoral fractures, intramedullary nailing, malunion, periaxial rotation, cone-beam CT, surface matching

I. INTRODUCTION

Locked intramedullary nailing (IMN) is the treatment of choice for comminuted fractures of the femoral shaft in the adult with union rates approaching 100%.1−4 Comminuted femoral fractures are clear breaks through the shaft with several splintered fragments populating the site. This type of injury is ideally suited to IMN where a long nail is passed through the femoral canal, bridging the fracture site. This nail is locked at both the proximal and distal bone fragments in order to prevent rotation, shortening or angulation at the fracture site. Techniques of indirect reduction and minimally invasive fracture fixation have been developed to minimize soft-tissue disruption at the fracture site. Unlike previous open techniques, these methods do not rely upon direct visualization of the fracture site, therefore anatomical reduction criteria such as correct limb length and proper alignment are more difficult to determine.5
The malalignment of a fractured femur consists of two distinct rotational components: an angulation of the long axis of the bone and a periaxial displacement that involves rotation about the bone axis. Angulation is readily apparent on radiographs and computed tomography (CT) as the displacement of the bone’s long axis. In contrast, periaxial rotation is more difficult to identify utilizing visual assessment of two-dimensional (2D) radiographs or three-dimensional (3D) scans with a limited field of view.6

Despite the popularity of femoral nailing, complications relating to poor alignment do occur with this minimally invasive technique.1,7,8 Angulatory and longitudinal malunion are carefully documented in most series.1,9,10 Rotational malunions, however, while addressed in some series,3,10–12 are often neglected.1,13,14 Periaxial malalignment is defined in the literature as a difference in alignment of more than 9°15 with respect to the contralateral femur. Furthermore, a study of 120 intramedullary femoral fracture reduction cases operated with the conventional technique found periaxial rotation malalignment of more than 15° in 19% of cases.16

Deformity secondary to malrotation can produce significant clinical symptoms, including pain and secondary degenerative joint disease, and can require surgical revision.16–20 Malrotation in patients after locked nailing is sometimes inferred from measurements of anteverision (inward twisting of the femur). Anteverision in normal populations varies tremendously, with a range of −4 to +36° reported by Hoaglund and Low.21 However, the difference in anteverision between sides in individuals is small, with an average difference of only 1° more anteverision in the right femur as compared to the left.22 While various techniques exist to assess femoral anteverision including CT, magnetic resonance imaging, and conventional radiographic techniques (teleradiography, orthoradiography), these techniques are not readily available in the operating room.23

Fracture reduction requires alignment of the two main fracture fragments along a longitudinal axis. With the introduction of a guide wire across the fracture site, bone angulation and out of axis displacement are set, and easily detected by fluoroscopy. Leg length and antetorsion are restored to the best extent possible by overlaying the contralateral femur on a fluoroscopic lateral view, but such methods often lack precision.23 This method also requires imaging of the contralateral leg, which is often draped-out intraoperatively, or the fracture may be bilateral, limiting availability for comparative examination. In simple fractures with clear fracture morphology, the alignment of the two fracture fragments is done under fluoroscopy as a puzzle matching of the fracture edges. This procedure is much more difficult when the fracture morphology is unclear due to comminution and/or bone loss. Ensuring intraoperative periaxial alignment of the femur remains a clinical challenge during minimally invasive fracture fixation. The current gold standard for determining the periaxial malrotation is postoperative CT, which requires additional surgical intervention if correction is required.24

Common clinical methods to determine periaxial rotation involve assessing the relative positions of several anatomical landmarks on x-ray or, on CT, measuring the angle between a reference line (e.g., anatomical plane) and a line defined by two or more landmarks on a transaxial slice.25 With CT images, a large field of view is required to identify landmarks and this is often problematic within the operating suite. Variation in slice angle can also introduce geometric distortions and quantification errors. Considering plane radiographs, bony anatomy is often obscured by superposition of adjacent structures and a relatively large field of view is also required to obtain sufficient landmarks. Furthermore, anatomic landmarks identified in radiographs and CT can be altered by age, disease, or trauma.

Several techniques based on robot-assisted reduction and surface matching have been developed for intraoperative use to reduce femoral shaft fractures.25,26 Gosling’s work on robotically manipulated proximal and distal fragments is still limited as the technique is based on visual perception related to the surgeon through two orthogonal cameras.26 Surface matching has been successful in realigning periaxial malrotation in chicken femora.25 The technique employs CT based unwrapped surface plots to register malrotated femora to a base line specimen. The authors demonstrated accuracy of the algorithm in registering malrotated femora to specimens originating from a different animal. This result demonstrates the strength of surface matching compared to anatomic landmark identification. This motivates further investigation of surface matching with respect to human long bones.

The mobile C arm has been used during trauma surgeries since its invention in the 1950s. Shapes and types of fractures, relative positions of fragments, implants, and surgical instruments can be visualized. C-arm technology is standard in intramedullary nailing of femoral shaft fractures, however, this technique has significant drawbacks such as a limited field of view and relating only two-dimensional information back to the surgeon.27

A new technology in intraoperative imaging28,29 is offered by the application of flat-panel detectors (originally developed for radiographic/fluoroscopic imaging) to cone-beam CT (CBCT) [Fig. 1(a)]. CBCT provides 3D volumetric image reconstructions from 2D projections acquired across a given source-detector trajectory about the patient. CBCT is
currently being utilized in pre-clinical systems under development for diagnostic and image-guided procedures and has been adapted to a mobile isocentric C-arm platform (Siemens PowerMobil) as a proving ground for application in image-guided surgery and interventional radiology. Similar in some respects to other C-arm-based 3D imaging systems (e.g., Iso-C-3D) there are important distinctions in this technology with respect to intraoperative guidance during orthopedic procedures. Intraoperative CBCT with a flat-panel detector demonstrates isotropic, sub-mm 3D spatial resolution and due to a larger field of view, the system was fixed to the distal femur fragment using a Schantz screw [Fig. 1(b)]. The surgeon aligned the proximal and distal femur and the position of the tool attached to the distal femur was recorded by the tracking system, thus setting the 0° rotation position. The femur was then wrapped with super-flab (a synthetic oil gel commonly used as a plastic bolus material in radiotherapy) to simulate the presence of soft tissue overlying the bone during imaging. The optical tracking measurements were used as truth to assess the periaxial malrotation.

C. Image acquisition

All images were acquired with a Siemens PowerMobil mobile, isocentric C arm, modified to include a flat-panel detector in place of the x-ray image intensifier, a servo drive for orbital motion, a method for geometric calibration, and a computer control system for image acquisition and 3D reconstruction. The system was operated in pulsed-fluoroscopic mode (1–6 p/s), with a magnification of 1.97 × and a field of view at the isocenter of 20 cm × 15 cm. The system uses a PaxScan 4030CB flat-panel detector (Varian Imaging Products, Palo Alto, CA) designed for real-time radiographic/fluoroscopic imaging, featuring a 2048 × 1536 pixel matrix (194 μm pixel pitch) in combination with a 0.6-mm-thick (~270 mg/cm²) CsI:Tl x-ray converter (Fig. 1). The panel can be read at frame rates up to 15 fps at full resolution, and up to 30 fps at half resolution (1024 × 768 pixels at 388 μm pitch). Cone-beam CT acquisition consisted of synchronized x-ray exposure and panel readout under continuous rotation of the C arm. A typical acquisition scheme was used consisting of 200 projections collected at 3.3 fps across the ~178° orbit (~60 s) at 100 kVp and 0.3 mA. The dose to isocenter of a 16 cm femur shaft was calculated to be 0.8 mGy for this technique, based on previously reported dose calculations and new exposure measurements for 2 mm Al and 0.1 mm Cu added filtration. Anecdotally, our work for head and neck scanning is done with CBCT acquisition at ~0.1–0.35 mSv compared to a nominal 2 mSv conventional diagnostic CT.

D. 3D volume reconstruction

Volume reconstructions were formed by 3D filtered backprojection using a Shepp-Logan filter, with mechanical flex of the C arm at different rotation angles accommodated by a geometric calibration. Two reconstructions were con-
considered in the study, a “low resolution” volume of dimension \((256 \times 256 \times 192)\) with voxel size 0.8 mm, and a “high resolution” volume of dimension \((512 \times 512 \times 384)\) with voxel size 0.4 mm. The increased spatial resolution achieved with the high resolution reconstruction comes at the expense of increased reconstruction time \((\sim 660 \text{ s vs } \sim 150 \text{ s})\) for high and low resolution, respectively) and larger data size \((192 \text{ vs } 24 \text{ MB})\). Reconstruction times were measured using the research system developed in-house for this study on a Dell Precision 650 PC [Dual 2.0 GHz Xeon CPU with 3 GB random access memory (RAM)], and can be reduced to \((\sim 80 \text{ s vs } \sim 17 \text{ s})\) using a high-speed, commercially available reconstruction platform (Exxim Cobra software).

E. Algorithm

A three-step algorithm was developed to quantify periaxial malrotation from the CBCT images (MATLAB, Fig. 2).

1. Bone surface extraction (Fig. 3)

The bone voxels are extracted from surrounding tissue in the scan using intensity thresholding. The scan is then cropped close to the cortical surface to reduce data size for improved processing speed. The center of the femur is defined on each axial slice as the centroid of all bone pixels. A polar grid is created around this center and the original axial slices are resampled onto radial coordinates. The bone surface of each slice is defined as the set of voxels furthest from the centroid over the set of sampled angles. The resulting data, \(BS(\theta, Z)\), is a matrix containing radius measurements for the bone surface surrounding the centroid, grouped for all slices. The data created by this process form a surface map of bone topology (Fig. 4).

To identify the proximal and distal ends of the bone, the algorithm detects sharp changes in voxel intensity along the length of the femur. For this procedure, the net voxel intensity for each slice is plotted as a function of slice number. The second derivative of these data is calculated and the local minima are associated with the ends of the scan and the midline of the femur, defined as the fracture site where a small physical gap exists. Using these three reference points, the scan data are separated into the proximal bone surface \([PB(\theta, Z_{\text{min to mid}})]\) and the distal bone surface \([DB(\theta, Z_{\text{mid to max}})]\).

2. Fracture gap extrapolation (Fig. 5)

To best smooth the proximal and distal bone surfaces for accurate extrapolation, two boxcar digital filters were tested:

![Fig. 2. One CBCT axial slice of the femur (a) and reconstructed isosurface of the malrotated specimen (b). Note the malalignment of the linea aspera.](image1)

![Fig. 4. Unwrapped surface plot of a femur as a function of distance from the femoral shaft centerline (a) and plotted with CT grayscale intensity values (b). Note the 30° malalignment of the linea aspera.](image2)
a one-dimensional filter with length 3 along the Z direction, and a two-dimensional square filter with length 5 along both the Z and \( \vartheta \) directions. To create a simulated fracture gap representative of comminution or bone loss, slices immediately above and below the fracture site were excluded from the proximal and distal bone surfaces. The remaining proximal and distal bone surfaces were then extrapolated through the simulated fracture gap to meet at the midline of the fracture in order to reconstitute the comminuted bone. A one-dimensional linear extrapolation was used along the longitudinal femoral axis at each sampled angle. The result is a proximal, PE(\( \vartheta \)), and distal, DE(\( \vartheta \)), extrapolated slice at the midline. To reduce noise and provide smooth signals for differentiation in Step 3, the extrapolated signals PE(\( \vartheta \)) and DE(\( \vartheta \)) were both filtered with a 20th order low pass FIR digital filter with normalized cutoff frequency \( w_n = 0.1 \).

3. Rotation optimization (Fig. 6)

Through data processing, proximal and distal data sets, PD(\( \vartheta \)) and DD(\( \vartheta \)), were calculated as functions of the proximal and distal extrapolations. To optimize this process, three different function routines were tested on PE(\( \vartheta \)) and DE(\( \vartheta \)): (1) first derivative, (2) second derivative and (3) raw data (no processing). Numerical approximations to the derivatives are computed from the set of differences between adjacent points in the data.

An estimate of the malrotation angle, \( \hat{\vartheta}_{\text{mal}} \), is then calculated as the argument that minimizes the vector norm between the rotated midline data DD(\( \vartheta + \hat{\vartheta}_{\text{shift}} \)) and the proximal midline data PD(\( \vartheta \)),

\[
\hat{\vartheta}_{\text{mal}} = \min_{\hat{\vartheta}_{\text{shift}}} \| DD(\vartheta + \hat{\vartheta}_{\text{shift}}) - PD(\vartheta) \|.
\]

The optimization of this objective function proceeds in two steps. First, a coarse estimate, \( \hat{\vartheta}_{\text{coarse}} \), is calculated as the argument that minimizes the objective function in Eq. (5) when \( \hat{\vartheta}_{\text{shift}} \) is restricted to a discrete grid between the lower and upper bounds, \( \vartheta_{\text{lb}} \) and \( \vartheta_{\text{ub}} \), with sampling interval \( \vartheta_{\text{samp}} \) degrees. Results presented in Sec. III are provided for \( \vartheta_{\text{lb}} = 75^\circ \) and \( \vartheta_{\text{samp}} = 1^\circ \). Second, \( \hat{\vartheta}_{\text{coarse}} \) is used as the starting point for a nonlinear least squares optimization with \( \hat{\vartheta}_{\text{shift}} \) defined over the continuous range bounded by \( \vartheta_{\text{lb}} \) and \( \vartheta_{\text{ub}} \). The motivation for this two-step process is to provide the nonlinear least squares function with a starting point sufficiently close to the global optimum to reduce the possibility of convergence to a local optimum.

F. Experimental protocol

The accuracy of \( \hat{\vartheta}_{\text{mal}} \) was tested for 16 femur random malrotations between \( -51^\circ \) and \( 60^\circ \) with simulated fracture gaps of 2, 6, 10, 20, 40, 60, 80 and 100 mm (Table 1). \( \hat{\vartheta}_{\text{mal}} \) values were compared to the optical tracking output, \( \vartheta_{\text{true}} \), at each of the 16 rotations with each of the nine simulated fracture gaps. Three malrotations were tested twice, during

<table>
<thead>
<tr>
<th>Gap (mm)</th>
<th>Slope (( \vartheta_{\text{true}} ) vs ( \hat{\vartheta}_{\text{mal}} ))</th>
<th>Pearson correlation (( r^2 ))</th>
<th>Correlation (Diff. vs Avg.)</th>
<th>Mean Diff. (( ^\circ ))</th>
<th>STDEV of Diff. (( ^\circ ))</th>
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the course of the accuracy experiment, to assess repeatability. Low and high resolution volumes were reconstructed to study the effect of reconstruction resolution on the algorithm. Finally, to assess the effect of scan alignment, three scans at 0° malrotation were imaged with the long axis of the femur axially aligned, at φ=10° angulation, and at φ=20° angulation to the scanning field. Pearson correlation coefficients, slopes as compared to the one-dimensional filter. The largest square boxcar digital filter yielded a lower error for all gap at a 6 mm gap compared to a 2.61°.

### III. RESULTS

#### A. Algorithm optimization

To smooth the raw bone surfaces, the two-dimensional square boxcar digital filter yielded a lower error for all gap sizes as compared to the one-dimensional filter. The largest average error for the 2D filter was 1.31° (±0.70°) observed at a 6 mm gap compared to a 2.61° (±1.93°) error for the one-dimensional (1D) filter. Furthermore, the 2D filter demonstrated functionality up to a gap of 60 mm, whereas the one-dimensional analysis was only accurate up to a 40 mm gap. The 2D filtered data were subsequently used in Steps 2 and 3 of the analysis.

Step 3 of the algorithm was optimized by testing different functions on the proximal and distal extrapolations. In choosing between the raw data, first and second derivatives for this optimization, the first derivative was the superior method. Whereas the first derivative method was functional up to a 100 mm gap, the raw data and second derivative analyses were only functional up to gaps of 60 mm (with considerable outliers resulting from the second derivative method). The first derivative method yielded an average error of only 1.18° (±0.78°) at 80 mm as compared to 40.09° (±51.08°) with the raw data and 5.81° (±15.72°) with the second derivative method.

Refinement of the algorithm included the utilization of a two-dimensional square boxcar digital filter for the surface data and running the vector norm minimization on the derivatives of the proximal and distal extrapolations. The subsequent analyses of low vs high resolution scans and the effect of specimen alignment relative to the scanning field were done utilizing these refined parameters.

#### B. Accuracy of femur rotation

Periaxial femoral malrotation was accurately calculated by the optimized surface matching algorithm compared to \( \hat{\theta}_{\text{true}} \) for gaps larger than 2 mm \( (r^2=1.00, \text{slope } 1.0) \) (Table I). The algorithm was unable to function at the gap size of 2 mm due to a physical gap in the specimen where CT slices were largely blank. The performance of the surface alignment was not impacted by gap sizes from 6 to 100 mm. This is of particular clinical significance since bone comminutions up to 100 mm in size would be ignored allowing for accurate reduction. Computational time for one scan was approximately 50 s (MATLAB V.7.0, Intel Pentium 4 1.6 GHz CPU with 1 GB RAM). The average difference between repeated measurements was 1.04° (over all gap sizes from 6 to 100 mm and three repeated malrotations). Worst-case repeatability between measurements was 2.07°, observed at a 100 mm gap and malrotation angle of −30.6°.

Using high-resolution CBCT data, accurate periaxial femoral malrotation was again very strongly predicted for gaps larger than 2 mm \( (r^2=1.00, \text{slope } 1.0) \) (Table I) and the performance of the surface alignment was not impacted by gap size from 6 to 100 mm. Computational time for one scan was approximately 220 s (MATLAB V.7.0, Intel Pentium 4 1.6 GHz CPU with 1 GB RAM). The average difference between repeated measurements was 1.22° (over gap sizes of 6–100 mm and three repeated malrotations). Worst-case repeatability between measurements was 3.91°, observed at a 60 mm gap and malrotation angle of 0.83°. Overall, only small differences were found between high and low resolution reconstructions. The use of low resolution reconstruction did not detract from the ability of the algorithm to accurately determine periaxial rotation (Fig. 7).

#### C. Oblique scanning

Due to the CBCT field of view, it was only possible to simulate fracture gaps up to 80 mm when scanning at an oblique angle to the specimen’s long axis. No trend was ob-

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**Table II. High-resolution results: high-resolution reconstruction comparison parameters for all fracture gaps (2–100 mm).**

<table>
<thead>
<tr>
<th>Gap (mm)</th>
<th>Slope ( (\hat{\theta}<em>{\text{true}} \text{ vs } \hat{\theta}</em>{\text{mal}}) )</th>
<th>Pearson correlation ( (r^2) )</th>
<th>Correlation (Diff. and Avg.)</th>
<th>Mean Diff. (°)</th>
<th>STDEV. of Diff. (°)</th>
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<td>0.26</td>
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<td>1.02</td>
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</table>
served with increasing fracture gap and little difference was found in the algorithm results due to angulation relative to the scanning field (Table III). Only at 20° of angulation with a 6 mm gap did errors rise, similar to findings at a 2 mm gap size with 0° angulation. Overall, periaxial malrotation can be accurately determined with angulations of the long bone axis at up to 20° to the scanning field.

IV. DISCUSSION

Rapid and accurate assessment of periaxial malrotation of femoral fractures was shown to be possible using surface data from single leg intraoperative CBCT images independent of specific anatomic landmarks. The angles of malrotation were accurately calculated ($r^2 = 1.00$, slope $\sim 1.0$) for 6–100 mm gaps. The image based surface matching algorithm was able to describe the periaxial malrotation with an accuracy of 1°.

Outliers were present only with the smallest 2 mm fracture gap. While extrapolation should be accurate at small gaps, the result is explained by the small physical gap present in the experimental specimen. This physical gap is the feature used to separate the proximal and distal segments. When processing a scan with only a 2 mm gap, the slices within the physical gap have very low voxel values, which propagate inaccuracies through the centroid calculation, resampling and extrapolation. While the algorithm was not notably affected by malalignment of the CBCT scan and the long bone axis of the femur, similar issues due to the presence of the physical gap in the scan resulted at a 6 mm gap for an oblique scanning angle of 20°. This limitation can be solved by ensuring that the slices above and below the fracture site are well within the bone by utilization of a minimum gap of 10 mm beyond the presence of the fracture gap on the CBCT images.

Low-resolution CBCT scans were processed considerably faster than high-resolution scans without any loss in accuracy. Neither speed nor radiation dose limit the clinical applicability of this technique.

Determining the correlation between the measurement differences and the average measurement ensures there is no dependency between these parameters. The generally low correlation values indicate that the difference does not get larger (or smaller) with increasing angle demonstrating that the error is due to random variation and not systematic due to the algorithm. Repeatability of the results was excellent due to the highly automated procedure that requires no specific landmark identification or subjective user inputs.

The accurate malrotation calculation can be explained by several interesting features of the algorithm. The resampling of the outer bone surface, BS($\hat{\theta}, Z$), about each slice centroid was able to account for bowing of the femoral axis. In further experimentation, this may also help to nullify effects of anatomic variation by customizing the resampling. In considering the filtering scheme, the 2D square boxcar filter was superior in smoothing the bone surface. The filtered data is a two-dimensional correlation of the original matrix as compared to the 1D digital filtering which only provided smoothing along the rows ($\hat{\theta}$).

Using the first derivative of the midline extrapolations minimized outliers and predicted malrotation with high accuracy. The raw data (no processing) and second derivative experimentations demonstrated local minima in the vector normalization. This caused occasional convergence onto the improper minimum and an inaccurate final answer, $\hat{\theta}_{\text{mal}}$. The first derivative of each midline function focused the vector minimization on the surface slope and accentuated the contour of the bone. This eliminated local minima and convergence was accurate for every trial.

Indirect fracture reduction remains a significant challenge in orthopaedic trauma. Minimally invasive surgery has been developed in order to spare soft tissue surrounding the fracture, however this approach reduces the exposure of the fracture and thus our ability to directly visualize the reduction. Intramedullary nailing is the treatment of choice for femoral fractures. While union rates are very high with intramedullary nailing, malrotations of the fracture reductions using fluoroscopic guidance are well documented. The integration of CBCT imaging and quantitative analysis presents an exciting advance aimed at minimizing femoral periaxial malrotation secondary to intramedullary nailing.

CBCT can provide rapid intraoperative high-resolution images with a large field of view (15 cm $\times$ 20 cm). Imaging data of this quality enables surface matching algorithms to be utilized even when large areas of severe comminution are present along the femoral shaft. The results ratified this novel method for assessing fracture rotation intraoperatively. The success of the surface matching algorithm in the femoral shaft motivates the development and application of this

<table>
<thead>
<tr>
<th>Angulation ($\phi^\circ$)</th>
<th>MeanDiff.</th>
<th>STDEV. of Diff.</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>1.76</td>
<td>1.16</td>
</tr>
<tr>
<td>10</td>
<td>2.02</td>
<td>0.91</td>
</tr>
<tr>
<td>20</td>
<td>2.10</td>
<td>1.40</td>
</tr>
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</table>

Table III. Oblique results: mean difference, ($\hat{\theta}_{\text{mal}}-\theta_{\text{true}}$), for increasing scan angulation of one specimen.
method to more complicated multiaxial fracture configurations and towards improving fracture alignment in other bones of the human skeleton.

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